

Two Methods for Analysis of Dynamic Scintigraphic Images of the Heart

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Abstract: *Dynamic scintigraphic images allow functional exploration of the considered organ and thus a good comprehension of the pathological phenomenon. There are two different approaches to analyse such images series; by following the activity of a region of interest or by computing parametric images. Both approaches may require the detection of an area of interest. This paper presents two methods of analysis of synchronised dynamic scintigraphic images of the heart based on an automatic detection of the left ventricle. The first method follows the time activity of the left ventricle and computes the ventricular ejection fraction. The second method computes a parametric image called "covariance image" where each pixel represents a covariance coefficient that indicates if the corresponding point in the heart and the reference area evolve or not in the same direction. The results of the application of the two methods on several dynamic images are presented and discussed.*

Keywords: *Image processing, segmentation, scintigraphic image, covariance, time activity curve.*

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1. Introduction

The medical imagery has been developed considerably these last years. At present, added to the conventional imagery, other methods allow obtaining various types of images: Echographic imagery, tomodensitometry (scanner X), Magnetic Resonance Imagery (MRI), and nuclear imagery. The latter provides not only anatomical and structural information but also quantitative and functional one. The three-dimensional imagery provides a 3D localization of the various organs and the probable existence of pathologies. Thus, the medical imagery constitutes an invaluable tool in the diagnostic of multiple pathologies.

Keeping pace with the development of medical imagery, the methods of image processing and analysis knew also a significant progress particularly in image segmentation. Thus the medical imagery, that represents for these methods an important application field, can computerize certain tasks that are generally performed manually which in return will improve its assistance to physicians.

The nuclear imagery is distinguished from the other methods of imagery by the fact that it allows a functional exploration of the considered organ and thus allows a good comprehension of the pathological phenomenon [13, 15, 16]. The acquisition of the image sequences allows learning more about the explored organ function. Visual analysis of these images can provide the physician with some information such as the presence or absence of anomalies [10] but such examination remains very limited. Image sequences should be processed by suitable techniques resulting from the field of image

processing to reveal information difficult to apprehend visually.

There are mainly two approaches of analysis of these sequences: The first one proceeds by "region of interest" and the second one is known under the name of "parametric imagery" [10]. The first approach consists in delimiting a "region of interest" in the considered organ and following its evolution in time which allows the estimation of the physiological functions such as activity-time, concentration-time, ventricular ejection fraction, etc. The delimitation of the area of interest boundary is generally done manually by the physician by means of an interactive device and sometimes only in the first image of the sequence. The same delimitation is applied by a simple copy transmitted to the other images. With this manual process, the results of investigation are different from one observer to another. Indeed, a study carried out by five different observers has given a variability between 19 and 43% [10]. So, it is important to use automatic methods to determine the area of interest based on techniques of image segmentation [10].

The second approach known under the name "of parametric imagery", considers each pixel an area of interest whose variation in time, named *dixel* (like "dynamic pixel") [10], represents a physiological function. For images of $N \times N$ resolution, there are N^2 physiological functions whose interpretation requires some transformations to extract one or several parameters. The matrix presentation of these parameters is called "parametric image". Factorial analysis [1, 3, 7, 8, 11] and Fourier analysis [7] are examples using this approach. The latter which is very

often used in nuclear angiocardigraphy, considers that the evolution of each pixel is a function of time and decomposes this 1-D function, into series of sines and cosines. The first harmonic provides two parametric images: Amplitude image and phase image whose visualization allows the physician to apprehend the existence of certain pathologies. The analysis in "principal components" [2] is another example of the parametric imagery. It aims to determine the main factors which are not correlated. In the case of the nuclear angiocardigraphy, one can be interested in the ventricular, or the auricular or the vascular factor. The determination of parametric image may also require automatic detection of an area of interest.

In this paper, we are interested in the design and implementation of two different methods to analyse synchronized scintigraphic image sequences of the heart. The first one proceeds by "area of interest" and the second one determines a "parametric image" representing a matrix of covariance. Both methods require the detection of the left ventricle which is made automatically by using image processing techniques. The second section of this paper presents the scintigraphic imagery and its characteristics. The third one presents the image processing steps that include a filtering operation and left ventricle segmentation. In the fourth and the fifth sections we will present the two analysis methods which have been applied to several sequences of scintigraphic images of the heart and their results have been compared to those obtained by the physician.

2. Scintigraphic Image

Nuclear imagery or scintigraphy is a science that allows the exploration of human body by injecting a slightly radioactive product and measuring its radiation during time. Because of its historical relationship with cancerology, it is often associated to the cancer detection and treatment. But it is also used in diagnosis, assessment of extension, monitoring and can play a role in the evaluation of the therapeutic response [13, 15, 16]. The radioactive product is associated and injected with a labelled molecule called a tracer that has a more or less specific tropism for a given organ in the human body. The more or less intense radiation emitted by the radioactive tracer is captured by a specific camera called "gamma camera" which was invented by Anger in 1958.

One of the components of this camera, known as "collimator", plays a significant role in the quality of the scintigraphic image. It appears as a wafer of lead or tungsten or another metal of high atomic number, bored with cylindrical or conical holes. Its task is to filter the photons gammas emitted by the source so that only those vertically emitted can reach the crystal. The number of photons passing through the collimator holes will be used to calculate pixel values of a digital

image. Thus, a three-dimensional piece of the observed organ is projected as a pixel of a 2D image. So, the value of each pixel represents the quantity of radioactive product included in an elementary volume of the observed organ rather than a colour. This constitutes one characteristic of the scintigraphic image.

The layout and the size of the collimator holes play an important role in the determination of the image quality. Indeed, the size of holes influences the image resolution and the detector sensitivity. With fine holes, their density and number can be increased and, consequently, the image resolution will be improved. On the other hand, the counting rate and the detector sensitivity are decreased. This requires an increase in the exposition time to have an acceptable image quality. Opposite effects are obtained with large holes. In general, the detector has a great sensitivity, but, because of the limited number of holes in the collimator and the low level of tracer radioactivity, scintigraphic image has a bad resolution and a bad quality that depends on the time of acquisition [16].

In spite of its bad resolution and bad quality, the scintigraphic image presents some characteristics which make it an invaluable tool of investigation and assistance to the diagnosis. Indeed, it allows to visualize in a simple way and at a low cost an inaccessible organ to conventional radiology, to carry out not only morphological images but also functional ones. This occurs because only the physiologically active zones collect the labelled molecule and the intensity of collection is related to the physiological or metabolic activity [16]. It allows the observation of the same amount of irradiation to the human body under several incidences and to follow the activity evolution of each area of the body. It also allows the acquisition of various types of images: Planar image, whole body image, synchronized image, and tomo-scintigraphic image. The continuous discoveries of new tracers ensure the nuclear imagery an endless potentiality [16].

The acquisition of sequences of scintigraphic images during time allows the observation of an organ activity. Therefore, one can follow the process of draining gastric or the activity of cardiac cavities. In angio-cardiography, as image acquisition requires the superposition of several images because of the weak counting rate, and as the heart morphology changes during its activity, it is important to superpose images taken at precise time of the cardiac cycle. So it is necessary to synchronize acquisition with Electrocardiogram signal (ECG). The computer connected to "Gamma camera" detects the wave R and divides the R-R interval into 16 equal segments during which 16 different images are captured. The acquisition takes around 500 cardiac cycles and each image of the sequence is the result of the superposition of the images captured in homologous intervals (Figure 1). Thus, each image of the sequence

represents the heart morphology at one specific time of the cardiac cycle [13].

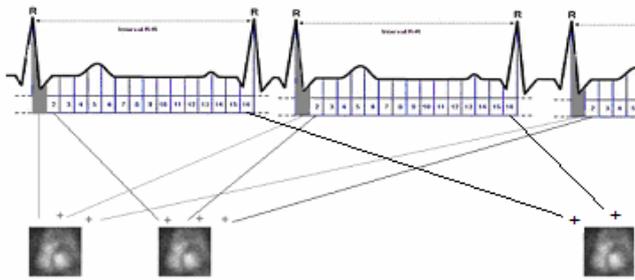


Figure 1. Synchronization of scintigraphic images acquisition with electro-cardiogram.

In this work we are interested in the process and the analysis of the synchronized scintigraphic images sequences which are acquired after labelling a tracer by technetium Tc99m. The objective is to computerize two methods of analysis of the cardiac activity: One by following the left ventricle activity and computing the ventricular ejection fraction and the other by determining and visualizing a parametric image representing the covariance function which is able to inform about the kinetics of the heart walls and thus to show the existence of certain pathologies. Both methods require the detection of the left ventricle.

3. Image Processing and Detection of the Left Ventricle

The bad contrast of scintigraphic image makes the cavities border difficult to locate even visually. Therefore, it would be necessary to process the image in order to reduce the quantity of noise which exists in the image and raise its contrast before the detection of the cardiac cavities. The flowchart presented in Figure 2, shows that the image undergoes mainly five operations: Filtering, enhancement, edge detection, filling and labelling and finally ventricle recognition.

3.1. Filtering

The scintigraphic image contains two types of noise: The first one resulting from the acquisition system and the second one, known as background noise, coming from the activity of the adjacent blood vessels to the cardiac cavities. In the absence of knowledge on the degradation model of the scintigraphic image, and knowing that our objective is to improve the image contrast, we choose to apply a median filter [4, 17, 20] by a 5x5 window. The feature of this nonlinear filter is to smooth the image while preserving contours. Other filters such as the average, the Gaussian, the minimum and the maximum filters have been tested but they did not give better results. The Gaussian filter does not smooth the homogeneous zones well. The maximum filter tends to increase the cavities with a risk of confusing the two ventricles. The minimum filter tends

to reduce them. Moreover, let us highlight that the larger the size of the filter is the more image is smoothed the more it loses its clearness.

3.2. Enhancement

The image undergoes then an operation of enhancement with an aim to improve its contrast. Knowing that the pixels values belong to a small range, an operation of contrast stretching [12] will enhance the image. The interval [a, b] of the transfer function (Figure 3) used to enhance the image, has been determined by analysing the image histogram. Figure 4 shows the results of image filtering and enhancement. In the enhanced image, the contrast between cardiac cavities and the background is improved.

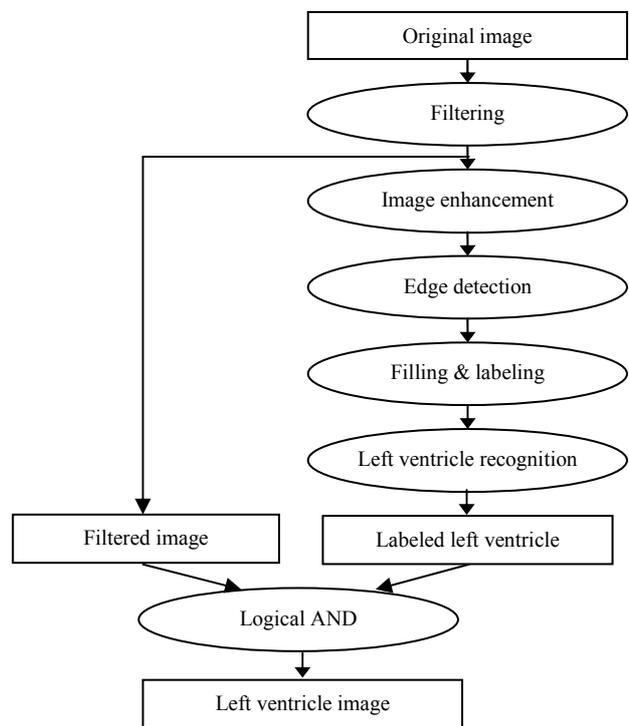


Figure 2. Image processing steps.

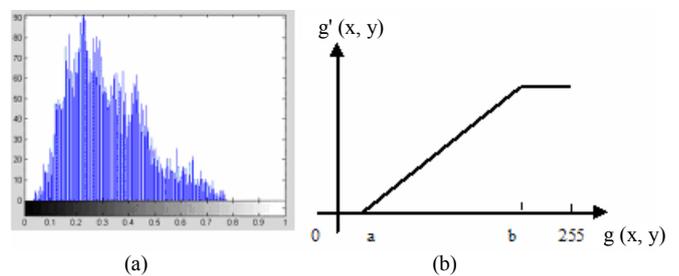


Figure 3. (a) Histogram of filtered image, (b) Transfer function for stretching.

3.3. Segmentation

There are mainly two dual approaches of image segmentation: The contour approach and the region approach. The first aims to determine the inside of the homogeneous regions whereas the second aims to detect their contours. The region approach requires the

definition of an homogeneity criterion which should be verified by pixels belonging to the same region. We have tested many methods using the derivative approach and advanced techniques such as fuzzy C-means classification [5], wavelet transformation combined with EM classification [14] and active contour [9]. But since the derivative approach gave us good results, we chose it for this task. Unfortunately, the bad contrast and the bad resolution of scintigraphic image make the cavities borders difficult to locate even visually. Thus, to prepare the detection of the left ventricle, it would be necessary to process the image in order to enhance the contrast between the cavities and the background. This process will affect only the segmentation result because the measurement will be done on the filtered image.

The application of a derivative method on the enhanced image allows the detection of the regions contours. The best result has been obtained by the zero cross of the Laplacian operator [17]. The sensitivity to noise of this operator requires its conjunction with a Gaussian filter. The connected components obtained (Figure 4-d) correspond to the cardiac cavities and a close vascular activity.

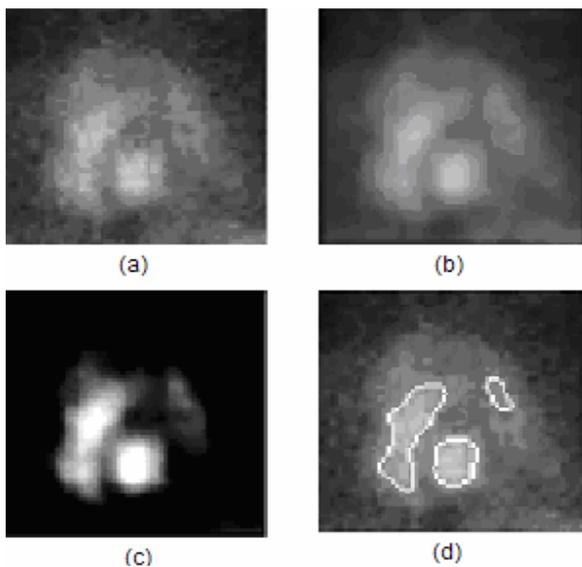


Figure 4. (a) Original image, (b) Filtered image, (c) Enhanced image, (d) Connected components with their borders.

3.4. Left Ventricle Recognition

The identification of the left ventricle is done after having carried out operations of filling, labelling and having applied a criterion using the rectangularity and surface descriptors on the connected components. A logical operation AND between the result and the filtered image selects the pixels belonging to the left ventricle (Figure 5).

4. First Method: Follow Up of the Left Ventricle Activity

This method proceeds by “region of interest” which is, in our case, the left ventricle. It aims to determine the activity-time curve and the ventricular ejection fraction which is a significant parameter for the physician. The two objectives require the estimation of the left ventricle volume.

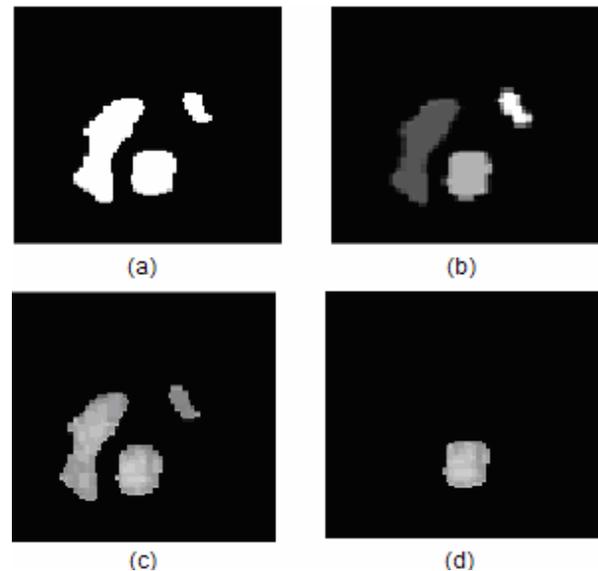


Figure 5. (a) Filled components, (b) Labeled components, (c) Components with their original pixels, (d) Left ventricle with its original pixels.

4.1. Estimation of the Left Ventricle Volume

In almost manual analysis of medical images, an object volume is estimated by a surface which is measured either by detecting a face contour of the object, or by approaching it by a geometrical model often more simple than the real form. Knowing that, in the case of the scintigraphic image, the value of a pixel represents the quantity of radiation emitted by an elementary volume, the sum of pixels belonging to the left ventricle represents its activity A . To estimate its volume V , it is necessary to subtract a background noise BN represented by the radiations emitted by the close vascular cavities [13].

$$V \approx A - BN \quad (1)$$

To estimate the background noise BN , the physician locates manually, by using an interactive device, an area close to the cardiac cavities. The sum of the radiations emitted by this area gives an estimation of the Background Noise (BN). This operation is done only in the first image of the sequence. The same value is used in the remaining images. In our method, this area is detected in each image automatically by the segmentation operation. Consequently, a different value of the background noise is computed in each image (Figure 5-c).

4.2. Computation of the Curve Activity-Time

The “curve activity-time” is defined by the evolution of the ventricular volume during a cardiac cycle. It informs about the functionality of the cardiac pump. Its shape shows either a normal kinetics or a presence of some types of pathologies such as hypokinesy and akinesy. The first one is due to a reduction in the amplitude of the parietal movement and the latter is due to a total absence of movement.

We have applied this method on eight sequences of images coming from healthy and sick subjects. Figure 6 presents two samples of time-activity curve coming from a healthy subject and a sick one. The curves are compared to those obtained by the physician. The average difference between the two curves is around 6%, but the most significant to highlight is that the shape of the curve is the same.

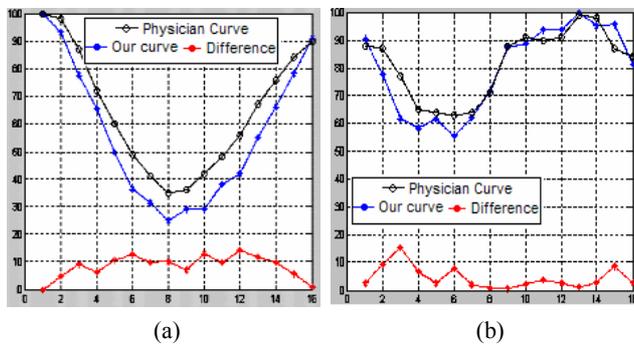


Figure 6. Samples of “curve activity time”, (a) Healthy case, (b) Pathologic case.

4.3. Ventricular Ejection Fraction

The Ventricular Ejection Fraction (VEF) is a parameter which informs about the faculty of blood ejection by the ventricle. It is defined by the following formula (2) [13]:

$$VEF = (V_{TD} - V_{TS}) / V_{TD} \quad (2)$$

Where:

V_{TD} and V_{TS} : Indicate respectively the diastolic volume (maximum filling of the heart) and systolic volume (minimal filling of the heart). By taking into account the equation (1), the formula (2) becomes:

$$VEF = ((A_{TD} - BN_{TD}) - (A_{TS} - BN_{TS})) / (A_{TD} - BN_{TD}) \quad (3)$$

Where:

A_{TD} and A_{TS} : Indicate respectively diastolic and systolic activity.

BN_{TD} and BN_{TS} : Indicate respectively diastolic and systolic background noise [13].

The computation of the ventricle ejection fraction is done according to the pseudo-code presented in the Figure 7.

Table 1 presents the results of a Ventricular Ejection Fraction compared to those obtained by the physician.

These results have been considered by the physician to be satisfactory.

- For each image of the sequence do
 - Process the image
 - Detect the left ventricle and the close vascular region
 - Compute the activity of the left ventricle
- Select the two images that correspond to the telediastole (maximum volume) and to the telesystole (minimum volume)
- Compute the quantities V_{TD} , V_{TS} , A_{TD} and A_{TS}
- Compute the VEF quantity

Figure 7. Pseudo-code for the computation of the ventricular ejection fraction.

Table 1. Results of ventricular ejection fraction.

Ventricular Ejection Fraction			
	Our Result	Physician Result	Difference
Sequence-1	68	64	4
Sequence-2	67.5	60	7.5
Sequence-3	67	60	7
Sequence-4	32.5	36	3.5
Sequence-5	60.5	65	4.5
Sequence-6	70	66	4
Sequence-7	58	55	3
Sequence-8	61.5	55	6.5

5. Second Method: Computation of the Covariance Image

The methods of parametric imagery consider that each pixel is an area of interest and are interested in the evolution of its activity in time. Thus there will be as many time-activity functions as pixels in the image. To have useful information, it is necessary to extract from these functions one or more parameters which one presents in the form of images known as parametric images or “functional images” [10]. Some methods are interested in the degree of similarity of pixel activity compared to the activity of a reference pixel or area suitably selected. For the determination of this degree of similarity, one can take as a reference the pattern recognition which deals with forms and is interested in their similarity. This similarity is measured by a distance, a correlation or a covariance [19].

In the case of a sequence of scintigraphic images, the form is a 1-D activity-time function of a pixel. The form of reference is the time-activity function of an area of the reference called Region Of Interest (ROI). The measurement of the similarity degree of these activities can inform the physician if the two areas move in the same direction or in the opposite one. The method used to measure this degree of resemblance is the covariance [6].

If we consider that the pixel activity is a random function $A(i, j, t)$, the average activity of an area of interest is also a random function $AR(t)$ given by the equation (4).

$$AR(t) = \frac{1}{M} \sum_{(i,j) \in RI} A(i, j, t) \quad (4)$$

Where M represents the number of pixels belonging to the region of interest.

The temporal statistical average of random functions $A(i, j, t)$ and of $AR(t)$ are respectively given by the equations (5) and (6).

$$\mu_A(i, j) = \frac{1}{T} \sum_{t=1}^T A(i, j, t) \quad (5)$$

$$\mu_{AR} = \frac{1}{T} \sum_{t=1}^T AR(t) \quad (6)$$

Where T represents the number of images in the sequence.

The covariance of two random variables is defined to be equal to the expected value of the multiplication of the differences between each random variable and its statistical average. Thus, the covariance of pixel (i, j) is given by equation (7) [6].

$$Cov(i, j) = \frac{1}{T} \sum_{t=1}^T (A(i, j, t) - \mu_A(i, j))(AR(t) - \mu_{AR}) \quad (7)$$

The sign of the covariance informs about the similarity degree of the activities of the two concerned areas. If it is equal to zero, the two activities are independent; if it is positive, they move in the same direction and if it is negative, they move in opposite one. Figure 8 shows the pseudo-code implemented to compute the covariance image.

- Determine the region of interest ROI
- For each image of the sequence do
 - Compute the average activity of the ROI: $AR(t)$
- Compute the temporal average μ_{AR}
- For each pixel (i, j) do
 - Compute its temporal activity $\mu_A(i, j)$
- For each pixel (i, j) do
 - Compute the covariance $Cov(i, j)$

Figure 8. Pseudo-code for the computation of the covariance image.

5.1. Choice of the Region of Interest

In nuclear angiocardiology, the physician would like to know if each area of the cardiac cavities functions correctly. In normal case, the two ventricles evolve in phase and the two auricles evolve in phase. But ventricles evolution and auricles evolution are in opposite phase. Therefore, the activity considered as a reference must be representative of a normal activity of one of these cavities. So, the region of interest must be selected inside a cardiac cavity for example in the left ventricle. Its size should be neither too large to ensure homogeneous pixels nor too small to mitigate the noise effect by taking the average of pixels activities.

The choice of the position of the region of interest is done in an automatic way in the first image of the sequence. After the location of the left ventricle, a circumscribed rectangle is determined in order to estimate the ventricle centre where a 3x3 region of interest is considered (Figure 9). This ensures that the region of interest will be always inside the ventricle during all the cardiac cycle. We have tested bigger sizes such as 5x5 and 7x7 but the best result has been obtained with a 3x3 region.

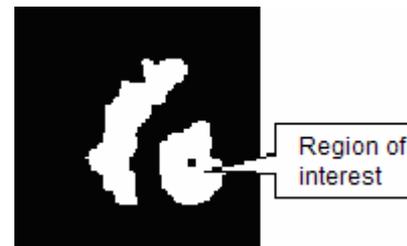


Figure 9. Position of the region of interest.

5.2. Tests and Results

We have tested the method on twenty sequences of real images resulting from healthy and sick subjects. For each sequence, the coefficients of the covariance of the various pixels are computed and presented in the form of a parametric image known as “covariance image” which, if it is displayed with a suitable look up colour table, shows the possible existence of pathologies. The results obtained are compared to the Fourier analysis ones that are computed by the workstation connected to the gamma camera. The physician has noted the conformity of the results except in the case of an akinesy pathology which corresponds to a total absence of movements of an area and thus gives a covariance close to zero. Figure 10 presents two samples of these results representing a healthy patient and a sick one having an hypokinesy pathology. Each result is compared to the phase image of Fourier analysis.

For a healthy subject, the two ventricles as well as the two auricles move in phase but the ventricles and the auricles move in opposite phase. This is confirmed by the covariance image in which the two ventricles are coloured in red and yellow corresponding to a positive covariance while auricles appear in blue representing a negative covariance. Fourier analysis shows similar results.

Dyskinesy is a paradoxical movement of an area of one of the two ventricles. That is due to the presence of a cicatricial area which provides, in a passive way, a pressure gradient effect and thus, functions in opposition of phase compared to its vicinity. The images presented in Figure 10 (c-d) correspond to a subject having a dyskinesy. Both Fourier analysis and covariance image confirm this pathology. The covariance image shows two areas within the left ventricle coloured differently in red and in blue which

correspond respectively to a positive and a negative value, indicating an opposition of phase within the same ventricle.

The results obtained by the application of "covariance image" method on 20 sequences have been compared to Fourier analysis ones and analysed by the physician. Table 2 which summarizes these results, shows that there is a conformity in 19 cases over 20. The 20th case presents two pathologies an akinesy and a dyskinesy. The covariance image doesn't show this pathology since the covariance value of an akinetic area is close to zero.

Table 2. Results obtained by the computation of covariance image on 20 cases.

	Number of Cases	Covariance Image	Fourier Analysis
Healthy cases	13	13	13
Hypokinetic cases	6	6	6
Cases having: - Dyskinesy - & akinetic	1	Dyskinesy	Dyskinesy & akinesy

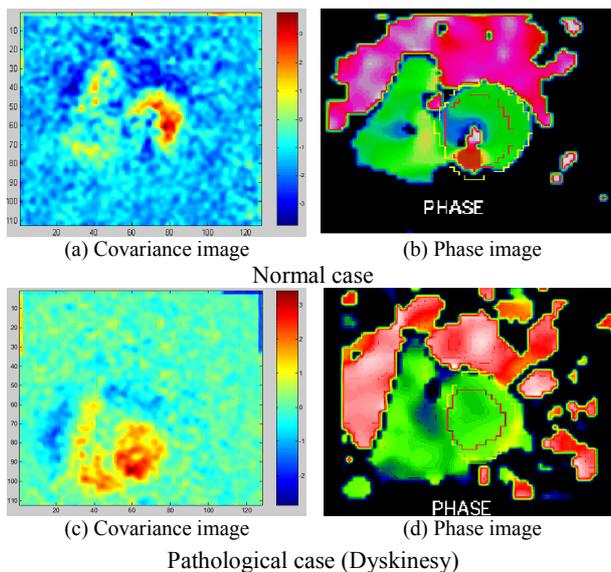


Figure 10 Samples of covariance image compared to phase image .

6. Conclusion

The scintigraphic image has the advantage of providing morphological and functional information. But it presents the disadvantage of having bad resolution and contrast. In nuclear angio-cardiography, the analysis of the dynamic scintigraphic images allows assistance in the diagnosis of certain pathologies. In the clinical routine, the automatic methods of analysis of such images use generally manual detection of the areas of interest. This manual procedure poses the problem of the variability of the results according to the observer. Image processing and analysis can provide advanced techniques in order to improve the image quality, locate regions of interest and show useful information.

In this paper, we have presented two methods for analysis of dynamic scintigraphic images. The first one detects the left ventricle, determines its activity-time and computes the ventricular ejection fraction. The second method detects the left ventricle, chooses a small reference area inside and computes a parametric image representing the covariance matrix. The two methods have been applied to several sequences of scintigraphic images of healthy and sick patients. The results of the first method were compared to those obtained by the physician who delimits manually the left ventricle. The curve activity-time presents a small variation, but its general form is similar and allows to make a conclusion if the patient has or hasn't a pathology. Regarding the ventricular ejection fraction the results are very close and remain in the conventional intervals. As for the second method, the analysis of the images of covariance allows the physician to diagnose both the normal and the pathological cases except for akinetic pathology which is difficult to see on the covariance image. The detection of the left ventricle which is a significant stage in the process of analysis can be improved by using more robust techniques against the presence of noise and the bad contrast.

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